LIMITED-ANGLE FREQUENCY-DISTANCE RESOLUTION RECOVERY IN NUCLEAR MEDICINE IMAGING

Background of the Invention

The present invention relates to the diagnostic imaging art.   It finds particular application in conjunction with nuclear single photon emission computed

5   tomography (SPECT) medical imaging and will be described with particular reference thereto. However, the invention will also find application in conjunction with other types of non-invasive diagnostic imaging.

Heretofore,   diagnostic   images   have   been

10   generated from single and multiple-head nuclear cameras. Typically, a patient positioned in an examination region is injected with a radiopharmaceutical. Heads of the nuclear camera are positioned closely adjacent to the patient to monitor the radiopharmaceutical. Typically,

15   the heads are stepped in increments of a few degrees around the patient until 360° of data have been acquired. That is, projection data along directions spanning 360° are collected. With multiple-head systems, projections along each direction need only be collected with one of

20   the heads and be assembled into a complete data set.

Each detector head carries a collimator which defines a path along which it can receive radiation. However, due to the finite length and dimensions of the collimator, each incremental area of the detector head

25   actually views an expanding cone. Thus, with increasing depth into the patient away from the detector head, the region from which a sensed radiation event originated expands.   This   creates   depth-dependent   blurring   and

uncertainty in the resultant image data. This error is a non-stationary convolution which is difficult to deconvolve. However, when the patient is viewed over a full 360°, the angular data sets are periodic in 2π

5 radians. By transforming the full data sets into the frequency domain with a Fourier transform fit to the sampling intervals, the non-stationary deconvolution is reduced to a stationary deconvolution problem, particularly for high frequencies.

10 Although these prior art resolution recovery techniques work well on full data sets, cardiac imaging is typically done using an incomplete data set. More specifically, in a three-head camera system where only two heads collect the emission data while the third head is

15 used to collect transmission data, the gantry is rotated by about 102° to generate the equivalent of only about 204° of emission data. The prior art resolution recovery techniques do not work on partial data sets whose data is not periodic in 2π radians.

20 The present invention contemplates a new and improved method and apparatus which overcomes the above-referenced problems and others.

Summary of the Invention

25 In accordance with one aspect of the present invention, a method of diagnostic imaging is disclosed. A plurality of projection data sets are collected at each of a plurality of angles around a subject. The projection images are collected over less than 360°. A resolution

30 recovery process is performed on the projection data sets. The resolution recovered projection data sets are reconstructed into an image representation.

In accordance with another aspect of the present invention, a method of diagnostic imaging is disclosed.

35 A gantry moves a detector head in a continuous angular orbit about a subject in an examination region. Data is collected during the continuous orbit and sorted into a

plurality of projection data sets corresponding to each of
a plurality of angular increments around a subject. A
resolution recovery process is performed on the projection
data sets. The resolution recovered projection data sets
5   are reconstructed into an image representation.

In accordance with yet another aspect of the
present invention, a diagnostic imaging apparatus is
disclosed. At least one detector head detects incident
radiation. A collimator mounted to the detector head
10   limits trajectories along which radiation is receivable by
the head. A movable gantry moves the detector head around
a subject in an examination region. A data acquisition
system acquires projection data sets from the detector
head at angular increments spanning less than 360°. A
15   zero-filling processor generates zero-filled data sets
between the actually collected projection data sets, to
create 360° of data sets. A smoothing processor smooths
interfaces between the actually collected and zero-filled
data sets. A resolution recovery processor operates on
20   the smoothed data sets. A reconstruction processor
reconstructs the resolution recovered data sets into a
three-dimensional image representation. An image memory
stores the three-dimensional image representation.

One advantage of the present invention is that
25   it accurately restores limited-angle data sets.

Another advantage of the present invention is
that it restores continuously scanned data sets.

Another advantage of the present invention is
that it processes image data in a clinically feasible
30   time.

Still further advantages and benefits of the
present invention will become apparent to those of
ordinary skill in the art upon reading and understanding
the following detailed description of the preferred
35   embodiments.

Brief Description of the Drawings

The invention may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only
5   for purposes of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 is a diagrammatic illustration of a nuclear medicine imaging system in accordance with the present invention;

10   FIGURE 2 is a diagrammatic illustration of the formation of the modified data set which is periodic in 2π radians, based on the non-periodic limited-angle data set;

FIGURE 3 shows the coordinate system used in data acquisition and image reconstruction.

Detailed Description of the Preferred Embodiments

15   With reference to FIGURE 1, a nuclear camera system 10 includes a plurality of detector heads 12, in the preferred embodiment three detector heads 12_1, 12_2, and 12_3. Two of the three heads are typically used to obtain
20   emission data, while the third head is used to obtain transmission data. Of course, other numbers of detector heads can also be utilized. Each of the detector heads includes a collimator 14_1, 14_2, and 14_3. In the preferred embodiment, the collimators collimate incoming radiation
25   from a subject 16 to parallel rays. However, because the collimators have finite size, each collimator permits rays which lie along a corresponding cone to pass to the detector head. The cone expands with depth into the patient from the detector head.

30   Each of the detector heads includes a reconstruction system 20_1, 20_2, and 20_3 which determines the coordinates on a face of the detector head in the longitudinal or z-direction of the patient and the transverse direction across the detector head. With the
35   detector head in a single orientation, scintillation events are collected for a preselected period of time to

generate a projection image representation. After the preselected data acquisition duration, a rotating gantry 22 rotates all three detector heads concurrently a short angular distance, e.g., 3°. In the new location, each of the detector heads collects another projection image. An angular orientation monitor 24 determines the angular orientation of each of the heads at each angular data collection position.

A data acquisition system 30 receives each of the projection images and an indication of the angle along which it had been taken. This data is stored in a three-dimensional memory 32 in longitudinal (z), lateral (s), and angular (ϕ) coordinates.

With continuing reference to FIGURE 1 and further reference to FIGURE 2, the rotating gantry 22 rotates the detector heads over 102° for cardiac imaging. Rotation over 102° generates the equivalent of about 204° of emission data 110 collected by two camera heads. Each data set is collected at equal angular increments, e.g., 3°. FIGURE 2 shows an incomplete data set with no data collected between 204° and 360° in the angular dimension. Simultaneously, the third camera collects transmission data over the 102° gantry rotation.

With continuing reference to FIGURE 2 and reference again to FIGURE 1, a resolution recovery system 42 includes an angular displacement dimension resolution recovery system and optionally includes resolution enhancement sub-system 46 for the axial dimension. The resolution recovery system in the angular direction (ϕ) has a zero-filling processor 50 which creates zero magnitude projection data sets 112 at each of the 3° intervals between 204° and 360°. In this manner, a function which is periodic in 2π radians is created. In order to prevent Gibbs' ringing, the sudden discontinuities at each end of the actually collected data 110 between the actually collected and zero-filled data 112 are smoothed by a smoothing processor 56. In the

preferred embodiment, the magnitude of the end points are each cut in half producing modified data points **114** and **116**. Optionally, other smoothing functions which span several points at each end are also contemplated.

A Fourier transform processor **60** transforms the data into the frequency domain. The Fourier transform is selected to match all of the sampling points including the zero-filled points in the angular dimension. A stationary deconvolution processor **62** operates on the frequency-spaced data. In frequency space, the deconvolution problem reduces to a stationary deconvolution problem, particularly for high frequencies. Because high frequencies correspond to fine detail, it is the high frequencies which correspond to the resolution to be optimized. The additional deconvolution **46** may optionally also be performed to improve image resolution in the z-direction. After the data has been deconvolved, an inverse Fourier transform processor **64** transforms the data from the frequency domain back into real space. Optionally, a three-dimensional memory **66** stores the resolution recovered sets of projection data. A reconstruction processor **68** reconstructs the projection data sets, using filtered back-projection, iterative reconstruction, or other techniques which are well-known in the art, and stores the resultant image in a three-dimensional image memory **70**. A video processor **72** withdraws selected portions of the reconstructed image and converts it to appropriate format for display on a human-readable monitor **74**, such as a video monitor, CCD display, active matrix, or the like. The video processor may withdraw selected slices, three-dimensional renderings, projection views, or the like.

In one alternate embodiment, the movable gantry **22** rotates the detector heads continuously. Although the detector heads rotate continuously, the data acquisition system **30** bins the collected data into regular angular intervals, e.g., 3°. Because the data is collected over

3° of rotation, there is an additional blurring component. As described in greater detail below, the stationary deconvolution processor **62** deconvolves the Fourier space data with respect to the blurring caused by the continuous
5   motion and treats data collected over a few degrees as if it were all collected at precisely the same angle.

      With reference to FIGURE 3, and looking now to details of the resolution recovery system, the working principles are explained in two-dimensions and circular
10   orbits, but the generalization to three-dimensions is straightforward and generalization to non-circular orbits is known to those conversed in the art.   The third dimension, z, is perpendicular to the plane defined by the x-y axes. The scanner collects data in coordinates (s, ϕ)
15   which are the sinogram coordinates of the Radon transform coordinates. The unblurred or undegraded Radon transform of the object o is the line integral along t, the axis perpendicular to s.   In a nuclear medicine tomographic device, the object o is blurred by a point response
20   function g. This process is modeled by a convolution in s. The amount of blurring depends on the depth t.   Hence, the blurring function is a non-stationary convolution of o with g.   The result is the blurred Radon transform p. For limited-angle tomography, this operation of the
25   scanner is represented by:

$$p(s,\phi) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} g(s-s',t)\, o(s'\theta+t\theta^{\perp})\, ds'\, dt \qquad (1),$$

where θ and $\theta^{\perp}$ are two-dimensional unit vectors aligned with the (s,t) axes:

$$\theta:[\cos\phi,\sin\phi] \quad \text{and} \quad \theta^{\perp}:[-\sin\phi,\cos\phi] \qquad (2).$$

A point source located at **x** in the object will have the point-source projection:

$$p_\delta = \int\limits_{-\infty}^{\infty}\int\limits_{-\infty}^{\infty} g(s-s',t)\,|J|\,\delta(s'-\mathbf{x}\cdot\theta, t-\mathbf{x}\cdot\theta^\perp)\,ds'dt = g(s-\mathbf{x}\cdot\theta, \mathbf{x}\cdot\theta^\perp) \quad (3),$$

where $|J|$ is the Jacobian of the change of coordinates from (x,y) to (s,t) and is equal to one because it is a rotation. The two-dimensional Fourier transform of Equation (3) gives:

$$\hat{p}_\delta(\omega,n) = \int\limits_{0}^{2\pi} e^{-in\phi}\int\limits_{-\infty}^{\infty} e^{-i\omega s} g(s-\mathbf{x}\cdot\theta, \mathbf{x}\cdot\theta^\perp)\,ds\,d\phi$$

$$= \int\limits_{0}^{2\pi} e^{-in\phi} e^{-i\omega\mathbf{x}\cdot\theta} \hat{g}(\omega,\mathbf{x}\cdot\theta^\perp)\,d\phi \approx \hat{g}\left(\omega, \mathbf{x}\cdot\theta^\perp = \frac{n}{\omega}\right)$$

$$(4).$$

5   The ϕ-integral is evaluated using the principle of stationary phase, an approximation process familiar to those knowledgeable in the art, which asymptotically converges to the exact value of the $\hat{p}_\delta$ integral at high frequencies. In Equation (4), it will be noted that the

10   solution does not depend on the location of the point source. Hence, it can be applied to any point source or a weighted collection of point sources.

Furthermore, if the scanning arc is limited to something less than 360° or 2π radians, the basic result

15   of stationary phase, that the distance t is given by:

$$t = \mathbf{x}\cdot\theta^\perp = \frac{n}{\omega} \quad (5)$$

is unchanged, because every distance is represented in each planar projection image. This enables the limits of integration in Equation (4) to be replaced with an arc or a series of disjoint arcs and Equation (5) still holds to

20   good approximation.

In operation, the digitized three-dimensional data set $p(s_j, z_k, \phi_m)$ is collected. The digitization assumes equal increments in each coordinate sample space. For example, the angular increment m is taken at intervals $\phi_m = 2\pi m/N$ radians, where N is an integer.

The depth dependent, point response function $g(s, z, t)$ is obtained. When the scan is over a limited angle arc described by the interval $[\phi_u, \phi_v]$, a digital input data set is formed according to the following process:

$$p_{in}(s_j, z_k, \phi_W) = \begin{cases} 0, & \text{for } 0 \leq W < \min(u, v) \\ \frac{1}{2}p(s_j, z_k, \phi_u), & \text{for } W = u \\ p(s_j, z_k, \phi_W) & \text{for } u < W < v \\ \frac{1}{2}p(s_j, z_k, \phi_v), & \text{for } W = v \\ 0, & \text{for } \max(u, v) < W < N \end{cases} \qquad (6).$$

The endpoint projection frames are multiplied by one-half to prevent excessive ringing or Gibbs' phenomena. Other smoothing functions for smoothing the sudden data discontinuity are also contemplated. The Fourier transform with respect to angle is selected to have dimension N, no more and no less, to preserve the cyclic nature of the data. The three-dimensional Fourier transform applied by the Fourier transform processor 60 of the modified projection data p_{in} is denoted by:

$$\hat{p}_{in}(\omega_s, \omega_z, n) \qquad (7).$$

The filter function:

$$\hat{g}\left(\omega_s, \omega_z, \frac{n}{\omega_s}\right) \qquad (8)$$

is defined. The filter function is used in a regularized inverse filter such as a noise reduction filter or a Metz filter to perform the stationary deconvolution 62. The

inverse Fourier transform processor **64** takes the inverse Fourier transform of the filtered image to obtain projection data that possesses improved resolution.

In the continuous scanning embodiment, the camera moves according to a preprogrammed orbit. In the preferred embodiment, the three heads move simultaneously in a complex patient dependent orbit. Continuous motion is advantageous for reducing motion complexity. However, continuous scanning degrades resolution. Preferably, the stationary deconvolution module includes a component to compensate for this degraded resolution.

Continuous scan projection data p_c is related to the step-and-shoot projection data **p** by:

$$p_c(s,z,\phi) = \frac{1}{\Delta\phi} \int_{-\Delta\phi/2}^{\Delta\phi/2} p(s,z,\phi')\,d\phi' \qquad (9).$$

The Fourier transform relationship is:

$$\hat{p}_c(\omega_s,\omega_z,n) = \hat{p}(\omega_s,\omega_z,n)\,\frac{\sin(n\Delta\phi/2)}{n\Delta\phi/2},\ -\frac{N}{2}+1 \leq n \leq \frac{N}{2} \qquad (10).$$

These relationships hold true for non-circular and limited-angle orbits as well. The filter function for continuous sampling $\hat{g}_c$, is modified to include the weighted sine function from Equation **(10)**:

$$\hat{g}_c\left(\omega_s,\omega_z,\frac{n}{\omega_s}\right) = \frac{\sin(n\Delta\phi/2)}{n\Delta\phi/2}\hat{g}\left(\omega_s,\omega_z,\frac{n}{\omega_s}\right) \qquad (11).$$

In accordance with another alternate embodiment, in the step-and-shoot imaging mode, the actually collected data integrals may be disjoint. It is not necessary for all of the angular views to be adjacent as in the example above.

The invention has been described with reference to the preferred embodiments. Obviously, modifications

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and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come

5   within the scope of the appended claims or the equivalents thereof.